(54) **IMPEDANCE CARDIOGRAPH APPARATUS**

VORRICHTUNG ZUR IMPEDANZ-KARDIOGRAPHIE

CARDIOGRAPHE A IMPEDANCE

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EP 0 771 172 B1

Description

Field of the Invention

[0001]   The present invention relates to impedance cardiographs which determine cardiac output by evaluating changes in impedance across the patient's chest cavity.

Background of the Invention

[0002]   Impedance cardiography is a non-invasive technique for determining cardiac performance in humans. When such equipment is employed, a high frequency electric signal is applied to the patient across outer electrodes positioned, for example, on the patient's head and lower thorax. Voltage differences between sensing inner electrodes positioned between the outer electrodes on the patient's neck and chest are measured and used to compute an impedance (Z). The impedance is based on the low magnitude, known electrical current passing between the outer electrodes.

[0003]   In 1932, Atzler and Leyman reported that cardiac output of a human could be determined by such impedance methods in *Uber ein neues Verfahren zur Darstellung der Herztatigkeit (Dielektrographie),* Arbeitsphysologie, 5: 636-680. In 1966, Kubichek reported the ability to correlate changes in base line impedance and the first derivative of impedance to stroke volume (SV) according to the following Equation (1) disclosed in U.S. Patent 3,340,867 and in the publication: *Development and Evaluation* of an *Impedance Cardiac Output System,* Aerospace Medicine, 37: 1208-1212.

$$SV = \frac{\rho * L^2 * \frac{\hat{dZ}}{dt} * L_{vet}}{Z_0^2} \tag{1}$$

where:

ρ is the resistivity of blood;
L is the spacing between the sensing electrodes;
Z_0 is an average or baseline impedance; and

$$\frac{\hat{dZ}}{dt}$$

is the magnitude of the peak negative value of the time derivative of the impedance Z for a period of time, typically a second.

[0004]   Cardiac output may be deduced from stroke volume by multiplying the latter times the heart rate.

[0005]   Although the Kubichek formula provides a value that correlated with cardiac output, the absolute accuracy of the method remained doubtful and, in particular, subjects with certain cardiovascular problems show values with great inaccuracies.

[0006]   In 1982, Sramek proposed a modification of the Kubichek formula of equation (1) which resolved the base line impedance Z_0^2 into a dynamic and static component as reported in the publication: *Cardiac Output by Electrical Impedance,* Med. Elect., 2:274-290. The static term Z_0s was described by the following equation:

$$Z_0s = \frac{\rho * L}{A} \tag{2}$$

where A is the area of the thorax being measured.

[0007]   The dynamic component Z_0d was simply the baseline or average of the impedance being measured:

$$Z_0d = Z_0 \tag{3}$$

[0008]   Incorporating the static term and dynamic term into the Kubichek equation provides the following formula:

$$SV = \frac{A * L * \frac{d\hat{Z}}{dt} * L_{vet}}{Z_0} \qquad (4)$$

[0009] The value of A may be estimated by approximating the chest as a cylinder in which case equation (4) becomes:

$$SV = \frac{C^2 * L * \frac{d\hat{Z}}{dt} * L_{vet}}{4 * \pi * Z_0} \qquad (5)$$

where C is circumference of the chest near the area of measurement. Alternatively, Sramek proposed that the term $\frac{C^2 * L}{4.25}$ be replaced with either $\frac{L^3}{4.25}$, or $\frac{(1.7 * H)^3}{4.25}$ where H is the height of the patient because 1.7*H approximates L.
[0010] These approximations did not produce good results and so Sramek was ultimately led to produce a set of charts attempting to establish correlation between area and the three factors of gender, height and weight.
[0011] In 1986, Bernstein proposed a modified equation in which the separation of the electrodes and the height of the patient were considered, in the following form:

$$SV = \frac{\left(\frac{L + 1.5 * H}{2}\right)^3 * \frac{d\hat{Z}}{dt} * L_{vet}}{4.25 * Z_0} \qquad (6)$$

[0012] All of the above methods suffer from lack of accuracy and indicate, in some subjects, falsely high or low values of stroke volume. According to a first aspect of the present invention there is provided an impedance cardiograph used to evaluate cardiac output of a human patient comprising:

means for applying an electrical excitation signal to the chest of a patient;
electrodes, adapted to be positioned on the patient for producing a first electrical signal Z which varies with impedance changes in the patient;
a non-linear processor for calculating a time derivative of Z, and providing a compressed derivative, having a value less than the time derivative of Z if the time derivative of Z is greater than a predetermined normal value, and having a value greater than the time derivative of Z if the time derivative of Z is less than the predetermined normal value; and
electrical circuit means for providing an indication of patient's cardiac stroke volume SV as a function of the compressed time derivative, cross-sectional area A of the patient's chest, and left ventricular ejection interval L_{vet}.

[0013] The present invention may thus provide an improved apparatus for deducing stroke volume (and hence cardiac output) from impedance measurements. The invention permits an improved estimation of body volume and a processing of the derivative of the impedance signal that improves the reliability of the derived values of stroke volume and cardiac output.
[0014] Preferably for a human patient having a height of H and a chest circumference C, an electrical excitation signal is applied to the patient as in conjunction with electrodes positioned on the chest with a separation distance of L, the electrodes producing a first electrical impedance signal Z which varies with impedance changes in the patient. A user input device, such as a keyboard, is provided to enter data on the height. Electrode separation distance and circumference to produce corresponding second and third electrical signals H, L and C indicating those values. An electrical circuit, which may be an electronic computer, receives the second and third electrical signal and provides an indication of the patient's cardiac stroke volume, SV, as a function of Z, A, L and left ventricular ejection interval L_{vet} where A is deduced by the following approximation:

$$A = \frac{L}{K * H * C}$$

where K is a predetermined constant.
[0015] Thus, the impedance cardiograph may calculate SV according to the following formula:

$$SV = \frac{L}{K*H*C} \; \frac{\rho * C^{2*} L * L_{vet}}{4*\pi*Z_0} \; \frac{d\hat{Z}'}{dt}$$

which may be simplified to:

$$SV = \frac{\rho * C^* L^2 * L_{vet}}{4*K*H*\pi*Z_0} \; \frac{d\hat{Z}'}{dt}$$

or by combining constants:

$$SV = \frac{\rho * C^* L^2 * L_{vet}}{K*H*Z_0} \; \frac{d\hat{Z}'}{dt}$$

[0016]   Thus, it is one object of the invention to provide a simple, yet more accurate characterization of a critical term A used in the calculation of stroke volume from patient impedance. In the present invention, the area of the impedance measurement, which is difficult to measure, is accurately derived from the readily measured values of chest circumference and patient height.

[0017]   The present invention has also recognized that variation in the derivative of impedance, one of the factors used in deducing stroke volume, is a significant source of inaccuracy in the computed stroke volume. Accordingly, whereas

$$\frac{d\hat{Z}}{dt}$$

may simply be the magnitude of the minimum derivative of impedance with time, it may also be compressed to reduce the amount that this maximum deviates from the norm. One method of weighting the maximum is according to the formula:

$$\frac{d\hat{Z}'}{dt} = \frac{d\hat{Z}n}{dt} = \frac{\dfrac{d\hat{Z}}{dt}}{\sqrt{\dfrac{d\hat{Z}}{dt} \Big/ \dfrac{\overline{\overline{dZ}}}{dt}}}$$

where:

$$\frac{d\hat{Z}}{dt}$$

is the magnitude of the minimum time derivative of Z;

$$\frac{d\hat{Z}n}{dt}$$

is the compressed value of

$$\frac{d\hat{Z}}{dt};$$

$\frac{\overline{dZ}}{dt}$ is a predetermined normal value and is an average value of

$$\frac{d\hat{Z}}{dt}$$

for a population.

[0018] Thus, it is yet another object of the invention to reduce the effect of a significant source of error in the calculation of stroke volume from impedance by implementation of a normal based weighting system.

[0019] The foregoing and other objects and advantages of the invention will appear from the following description. In the description, reference is made to the accompanying drawings which form a part hereof and in which there is shown by way of illustration, a preferred embodiment of the invention. Such embodiment does not necessarily represent the full scope of the invention, however, and reference must be made therefore to the claims herein for interpreting the scope of the invention.

Brief Description of the Drawings

[0020]

Fig. 1 is a block diagram of the circuitry for an impedance cardiograph according to the present invention, showing a computer as used to analyze the impedance data to produce a value of stroke volume and cardiac output; and

Fig. 2 is a flow chart of the software employed by the computer of Fig. 1 in analyzing the impedance and ECG signals acquired from the patient.

Detailed Description of the Preferred Embodiment

[0021] Referring to Fig. 1, a portable impedance cardiograph 10 is connected to a patient 12 by five patch electrodes 14(a)-(e). The first electrode 14(a) may be positioned on the patient's skin behind the right ear at the level of the ear canal. The second electrode 14(b) may be located at the left side of the neck on a flat surface approximately between the level of the chin and base of the hairline at least 5 cm below the level of electrode 14(a). The third electrode 14(c) may be located just above the base of the sternum on the anterior median lines. The fourth and fifth electrodes 14(d) and 14(e) may be located at least 5 cm below the electrode 14(c) on the patient's right and left sides respectively at the costal arch and the anterior axillary line.

[0022] Electrodes 14(a) and (d) are attached to an oscillator 18 which produces a constant current of approximately 1 milliamperes RMS through the patient 12. This electrical excitation establishes a series of equal potential surfaces through the patient 12 perpendicular to a line extending between the two outer electrodes 14(a) and 14(d).

[0023] Electrodes 14(b) and 14(c) may sense the equal potential lines generated by the current flowing between the outer electrodes 14(a) and 14(d). Because the current between the outer electrodes 14(a) and 14(d) is of constant amplitude, the amplitude of the voltage sensed between the inner electrodes 14(b) and 14(c) is proportional to the thoracic impedance of the patient 12. The inner electrodes 14(b) and 14(c) are connected to differential amplifier 20 producing a signal Z.

[0024] The amplifier 20 includes isolation circuitry that electrically isolates the inner electrodes 14(b) and 14(c) from the subsequent circuitry. Differential amplifier 20 also includes a precision half wave rectifier and low pass filter so as to provide a slowly varying DC signal whose value is proportional to the impedance being measured. The input impedance of the differential amplifier 20 is very high (e.g. 10 megohms) as compared to the impedance of the patient 12 between the inner electrodes 14(b) and 14(c). Thus, negligible current will flow through the inner electrodes 14(b) and 14(c) to amplifier 20.

[0025] The impedance signal Z is received by a multiplexer 32, such as are known in the art, to be periodically connected to an analog to digital converter 22 which samples the signal and provides a binary data word that may be read by microprocessor 24 via a bus 26.

[0026] A first, vertical ECG signal is measured across electrodes 14(b) and 14(c) lying generally along a generally vertical line. This ECG signal is received by differential amplifier 30 to produce an electrocardiograph signal E_v according to techniques well known in the art.

[0027] A second, horizontal ECG signal is measured across electrodes 14(c) and 14(e) lying generally along a horizontal line. This ECG signal is received by differential amplifier 31 to produce an electrocardiograph signal E_h. Both signals E_v and E_h are received by multiplexer 32 which periodically connects these signals to the analog to digital converter 22 to be sampled and converted to digital words for transmission on internal bus 26.

[0028] Differential amplifiers 30 and 31 receiving the ECG signal also include isolation circuitry and a low pass filter having a cut-off frequency such as to substantially remove the 50 kHz oscillator signal from oscillator 18. As will be described further below, the direct and quadrature ECG signals are combined to produce a single ECG signal largely independent of the electrical orientation of the patient's heart.

[0029] Also attached to bus 26 is computer memory 34 which may be composed of both random access memory ("RAM") and read only memory ("ROM") according to well known computer architectures. Memory 34 provides a means for storage of the binary representations of signals Z and E_v and E_h under the control of microprocessor 24, and also holds a stored program defining the operation of the microprocessor 24 for the calculation of cardiac output as will be

described.

[0030] Also attached to bus 26 is a pushbutton switch 25 which may be used by the patient to mark the occurrence of some event, such as a cardiac episode, during the recording of data from the patient 12 as will be described.

[0031] The bus 26 also communicates with an infra-red transceiver 36, which permits the microprocessor to transmit and receive data to and from a similar transceiver 35 in a detachable base unit 37. The transceiver 35 is connected by a serial cable 38 to a desk-top computer 42 having a video monitor 43, a keyboard 44 and a disk drive 48 such as are known in the art. As such the impedance cardiograph 10 is portable and may be powered by internal batteries so as to be carried with the patient in the manner of a Holter monitor.

[0032] Generally, the impedance cardiograph receives signals from the patient 12, isolates, amplifies and filters those signals, and then translates the signals to digital values which may be read and stored by microprocessor 24 to be processed according to the stored program in memory 34. Results of the processing may be transmitted to the computer 42 to be displayed on the video monitor 43 or saved on disk 48.

[0033] The operation of the impedance cardiograph 10 according to the stored program is controlled by a human operator through keyboard 44. The operator prepares the patient 12 for the impedance measurements and may enter certain data to the keyboard 44 that characterizes the patient 12 and that is necessary for the analysis of the signals from the patients 12 as will be described. This analyses is done in partially in the microprocessor 24 so as to reduce the amount of data to be stored in memory 34, but, as will be understood in the art, the analyses of the data may be shared between the microprocessor 24 and the computer 42 as a matter of engineering choice.

[0034] Referring now to Fig. 2, at the first step in the analyses, indicated by process block 50, certain data related to the particular patient 12 or related to fundamental and essentially universal physiological parameters, may be entered by the operator.

[0035] The first of these parameters is ρ which is the resistivity of blood in ohms-cm. Generally, this value may be approximated as a constant for all patients, however, it may be modified by the operator in abnormal cases based on the measurement of hematacrit.

[0036] A second value, C, is the circumference of the patient's chest in cm taken at the site of inner electrode 14(c) around the patient's chest.

[0037] The value L is also input, being the distance between the inner electrodes 14(b) and 14(c) in cm. The height of the patient, H, in cm is also entered.

[0038] Once the necessary fixed parameters are entered at process block 50, two variable parameters: heart rate HR and ventricular ejection time L_{vet} are entered per process block 52. For this purpose, the ECG signal E may be directly displayed on the video monitor 43 so that these quantities may be determined according to methods well known in the art. The ECG signal is calculated from the vector sum of the values of E_v and E_h most simply as follows:

$$ECG = \sqrt{E_v^2 + E_h^2}$$

[0039] This vector summing reduces the need to precisely orient the ECG electrodes with respect to an electrical polarity of the heart and therefore in practice provides a superior ECG signal.

[0040] Generally, L_{vet} is the time between the opening of the aortic valve and the closing of the aortic valve. The heart rate is simply the number of beats per second which is the inverse of the period between successive R waves. The heart rate may be averaged over a number of beats according to methods well known in the art. Both quantities may be determined by inspection by the operator or preferably may be determined automatically after sufficient ECG and Z data is acquired as indicated by process block 52. The value of L_{vet} in the preferred embodiment is determined by analyzing the impedance signal Z to measure a period beginning when $\frac{dZ}{dt}$ is first less than zero and ending when the value of $\frac{dZ}{dt}$ reaches a local maximum above zero. The heart rate HR is measured by detecting and counting R waves in the ECG signal.

[0041] The acquisition of the ECG and Z signal per process block 54 continues. This acquisition is on a continuous basis and occurs concurrently with the subsequent calculations so that the cardiac output may be continuously stored in memory 34 or displayed in essentially a real-time manner.

[0042] The acquired impedance data is in the form of discreet samples taken approximately 300 times per second, each sample which may be represented by Z_i where i is an index number of the particular sample. As each sample Z_i is acquired, it is stored in consecutive addresses in memory 34 to indicate its relative position with respect to other samples and to indicate the time of the sample indirectly through the constant sampling rate.

[0043] Because the impedance cardiograph 10 is portable, there is a risk that artifacts may be introduced into the impedance measurement by electrode movement. This is because the measured impedance values are approximately two orders of magnitude lower than the electrode to skin resistance. Accordingly, the impedance data over a period of approximately one minute is "ensemble" averaged, per process block 56, to reduce its noise content. Ensemble averaging is a well known technique in which blocks of impedance data are averaged on a point-by-point basis with other

blocks of impedance data so that the averaged points are from corresponding portions of the impedance waveform cycle. Thus, the shape of the impedance waveform is not destroyed in the averaging process. In order to perform such ensemble averaging, it is necessary to identify a common fiducial point to align the blocks of data. Selection of the fiducial point must be extremely precise, otherwise the characteristics of the impedance waveform will be "blurred" by a misregistration of other blocks.

[0044] This fiducial point may be the peak of the R wave of the ECG signal. Normal techniques for determining the time of the R wave, such as may be used for the measurement of heart rate, however, are not suitably accurate for the purpose of ensemble averaging. Accordingly an extremely accurate identification process is used. First, as represented by process block 58, the ECG signal is monitored to isolate a standard R wave. Only portions of the received ECG signal having no detectable artifacts or noise are considered. This standard R wave is then correlated to the incoming ECG signal to identify the precise location of the R wave (by the value of highest correlation). This location is used at the point of common alignment for the impedance waveforms to be ensemble averaged. Periodically, a new standard R wave is obtained so that the standard remains current over time.

[0045] After pairs of data Z_i and Z_i+1 are acquired and averaged, a derivative value is $\frac{dZ_i}{dt}$ may be computed by a simple subtraction of adjacent samples of the ensemble average per process block 58, that is:

$$\frac{dZ_i}{dt} = Z_{i+1} - Z_i \tag{7}$$

[0046] Alternatively, in order to reduce the presence of 50 Hz or 60 Hz noise, this derivative computation can employ samples $Z_{i+6} - Z_i$ or Z_{i+5}, respectively.

[0047] It has been determined that variation in the magnitude of the minimum of this value,

$$\frac{d\hat{Z}}{dt}$$

, is a significant source of error in the calculation of cardiac output. Accordingly, at process block 60, a compressed derivative,

$$\frac{d\hat{Z}n}{dt}$$

is computed according to the following formula:

$$\frac{d\hat{Z}n}{dt} = \frac{\frac{d\hat{Z}}{dt}}{\sqrt{\frac{d\hat{Z}}{dt}/\overline{\frac{d\hat{Z}}{dt}}}} \tag{8}$$

where:

$$\frac{d\hat{Z}}{dt}$$

is the magnitude of the minimum time derivative of Z as previously defined;

$$\frac{d\hat{Z}n}{dt}$$

is the compressed value of

$$\frac{d\hat{Z}}{dt};$$

$\overline{\frac{dZ}{dt}}$ is the predetermined normal maximum value and is an average value

$$\frac{d\hat{Z}}{dt}$$

for a population and is about 1.73 ohms per second.

[0048] Equation (8) has the effect of reducing the excursions of

$$\frac{d\hat{Z}}{dt}$$

from $\frac{d\overline{Z}}{dt}$ Other nonlinear compression systems may also be used provided they have the effect of compressing

$$\frac{d\hat{Z}}{dt}$$

about the norm.

[0049] At succeeding process block 52, the value of Z_0 is also computed. Z_0 is the base transthoracic impedance and in this implementation, simply the average value of Z for one cardiac cycle. Because of the need to average a number of samples, when the sampling is first begun, no display is provided to the video terminal 43 until sufficient samples have been made to insure the accuracy of this value Z_0.

[0050] At process block 64 a stroke volume may be calculated according to the following formula:

$$SV = \frac{L}{K*H*C} \; \frac{\rho*C^{2}*L*L_{vet}}{4*\pi*Z_0} \; \frac{d\hat{Z}}{dt} \tag{9}$$

[0051] It will be recognized that this is simply Equation (5) with the addition of a factor ρ and the addition of a factor $\frac{L}{K*H*C}$ this latter factor approximating a slice of body volume in the area of the impedance measurement.

[0052] This calculation of stroke volume has shown significant improvements in accuracy and correlation coefficients in clinical studies.

[0053] Also at process block 64, cardiac output may be determined by multiplying the stroke volume times the heart rate:

$$CO = SV*HR \tag{10}$$

[0054] As cardiac output is computed, it is displayed in graphical form on video monitor 43. Thus, the operator is provided with concurrent ECG data and cardiac output data on an essentially real-time basis per the display indicated by process block 66.

[0055] After each updating of the display of 43 is accomplished, the program acquire additional data until the measurement session is complete.

[0056] While this invention has been described with reference to particular embodiments and examples, other modifications and variations will occur to those skilled in the art in view of the above teachings. Accordingly, the present invention is not limited to the preferred embodiment described herein, but is instead defined in the following claims.

Claims

1. An impedance cardiograph (10) used to evaluate cardiac output of a human patient (12) comprising:

   means (14a,14b,18) for applying an electrical excitation signal to the chest of a patient;
   electrodes (14c,14d), adapted to be positioned on the patient (12) for producing a first electrical signal Z which varies with impedance changes in the patient;
   a non-linear processor (24) for calculating (59) a time derivative of Z, and providing (60) a compressed derivative, having a value less than the time derivative of Z if the time derivative of Z is greater than a predetermined normal value, and having a value greater than the time derivative of Z if the time derivative of Z is less than the predetermined normal value; and
   electrical circuit means (42) for providing an indication of patient's cardiac stroke volume SV as a function of

the compressed time derivative, cross-sectional area A of the patient's chest, and left ventricular ejection interval L_{vet}.

2. The impedance cardiograph of claim 1, wherein the non-linear processor (24) produces the compressed value of the time derivative of Z according to the following formula:

$$\frac{\hat{dZn}}{dt} = \frac{\frac{\hat{dZ}}{dt}}{\sqrt{\frac{dZ}{dt} / \frac{\hat{dZ}}{dt}}}$$

where:

$$\frac{\hat{dZ}}{dt}$$

is a time derivative of Z,

$$\frac{\hat{dZn}}{dt}$$

is the compressed time derivative of Z, and

$\frac{dZ}{dt}$ is the predetermined normal value and is an average value of

$$\frac{\hat{dZ}}{dt}$$

for a population.

3. The impedance cardiograph of claim 1, wherein the predetermined normal value is substantially 1.73 ohms per second.

4. The impedance cardiograph of claim 1, wherein:

the electrodes (14c,14d) are adapted to be positioned on the patient with a separation distance of L; and further comprising an input device for receiving (50) values of the height and circumference and providing corresponding second and third electrical signals H and C indicating those values;

wherein the electrical circuit means (24) receives the first, second and third electrical signals and provides the indication of the patient's cardiac stroke volume SV as a function of Z, L, and deduces A by the following approximation:

$$\frac{L}{K*H*C}$$

where K is a predetermined constant.

5. The impedance cardiograph of claim 4, wherein the electric circuit means (24) calculates SV according to the following formula:

$$SV = \frac{L}{K*H*C} \frac{\rho*C^2*L*L_{vet}}{4*\pi*Z_0} \frac{\hat{dZ_n}}{dt}$$

where:

ρ   is the resistivity of blood;

Z_0 is an average baseline impedance;

L_{vet} is a left ventricular ejection interval; and

$$\frac{\hat{dZ_n}}{dt}$$

is the compressed time derivative of the impedance Z.

6. The impedance cardiograph of claim 1, wherein

$$\frac{\hat{dZ_n}}{dt}$$

is the magnitude of the minimum compressed time derivative of Z.

7. The impedance cardiograph of claim 4, wherein K is substantially 0.875.

Patentansprüche

1. Impedanz-Kardiograph (10), der verwendet wird, um das Herzminutenvolumen eines Menschen (12) zu bewerten, umfassend:

    Mittel (14a, 14b, 18) zum Anlegen eines elektrischen Anregungssignals an die Brust eines Patienten;

    Elektroden (14c, 14d), die zur Anordnung auf dem Patienten (12) ausgebildet sind, um ein erstes elektrisches Signal Z zu erzeugen, das mit Impedanzänderungen im Patienten variiert;

    einen nichtlinearen Prozessor (24) zum Berechnen (59) einer Zeitableitung von Z und Bereitstellen (60) einer komprimierten Ableitung, deren Wert geringer als jener der Zeitableitung von Z ist, wenn die Zeitableitung von Z über einem vorbestimmten Normalwert liegt, und deren Wert größer als die Zeitableitung von Z ist, wenn die Zeitableitung von Z unter dem vorbestimmten Normalwert liegt; und

    Schaltkreismittel (42) zum Bereitstellen einer Anzeige des Herzschlagvolumens SV eines Patienten als Funktion der komprimierten Zeitableitung, der Querschnittsfläche A der Brust des Patienten und des Ausstoßintervalls der linken Herzkammer L_{vet}.

2. Impedanz-Kardiograph nach Anspruch 1, worin der nichtlineare Prozessor (24) einen komprimierten Wert der Zeitableitung von Z nach der folgenden Formel erzeugt:

$$\frac{\hat{dZ_n}}{dt} = \frac{\frac{\hat{dZ}}{dt}}{\sqrt{\frac{dZ}{dt} / \frac{\hat{dZ}}{dt}}}$$

worin

$$\frac{\hat{dZ}}{dt}$$

    eine Zeitableitung von Z ist,

10

$$\frac{\hat{dZ_n}}{dt}$$

5   die komprimierte Zeitableitung von Z ist, und

$\frac{dZ}{dt}$ der vorbestimmte Normalwert ist und ein Mittelwert von

10

$$\frac{\hat{dZ}}{dt}$$

für eine Population ist.

3. Impedanz-Kardiograph nach Anspruch 1, worin der vorbestimmte Normalwert im Wesentlichen 1,73 Ohm/s ist.

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4. Impedanz-Kardiograph nach Anspruch 1, worin:

die Elektroden (14c, 14d) dazu ausgebildet sind, am Patienten in einem Trennabstand L angeordnet zu sein; und

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die weiters eine Eingabevorrichtung zum Empfangen (50) von Werten der Höhe und des Umfangs und Bereitstellen entsprechender zweiter und dritter elektrischer Signale H und C umfasst;

worin das Schaltkreismittel (24) das erste, das zweite und das dritte elektrische Signal empfängt und die Angabe
25   des Herschlagvolumens SV des Patienten als Funktion von Z, L liefert und A durch die folgende Annäherung ableitet:

$$\frac{L}{K*H*C}$$

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worin K eine vorbestimmte Konstante ist.

5. Impedanz-Kardiograph nach Anspruch 4, worin das Schaltkreismittel (24) SV nach der folgenden Formel berechnet:

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$$SV = \frac{L}{K*H*C} \frac{\rho*C^2*L*L_{vet}}{4*\pi*Z_0} \frac{\hat{dZ_n}}{dt}$$

40   worin

     ρ der spezifische elektrische Widerstand von Blut ist;
     Z_0 eine mittlere Basislinienimpedanz ist;
     L_{vet} ein Ausstoßintervall der linken Herzkammer ist; und

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$$\frac{\hat{dZ}}{dt}$$

die komprimierte Zeitableitung der Impedanz Z ist.

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6. Impedanz-Kardiograph nach Anspruch 1, worin

$$\frac{\hat{dZ_n}}{dt}$$

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die Größe der minimalen komprimierten Zeitableitung von Z ist.

7. Impedanz-Kardiograph nach Anspruch 4, worin K im Wesentlichen 0,875 ist.

Revendications

1. Cardiographe à impédance (1) utilisé pour évaluer le débit cardiaque d'un patient humain (12) comprenant :

   des moyens (14a, 14b, 18) pour appliquer un signal d'excitation au thorax d'un patient ;
   des électrodes (14c, 14d), adaptées pour être positionnées sur le patient (12) en vue de produire un premier signal électrique Z variant avec les changements d'impédance chez le patient ;
   un processeur non linéaire (24) pour le calcul (59) d'une dérivée par rapport au temps de Z, fournissant (60) une dérivée compressée, ayant une valeur inférieure à la dérivée par rapport au temps de Z si la dérivée par rapport au temps de Z est supérieure à une valeur normale prédéterminée, et ayant une valeur supérieure à la dérivée par rapport au temps de Z si la dérivée par rapport au temps de Z est inférieure à la valeur normale prédéterminée ; et
   un moyen de circuit électrique (42) destiné à fournir une indication du débit systolique SV du patient comme fonction de la dérivée par rapport au temps compressée, l'aire transversale A du thorax du patient, et de l'intervalle d'éjection ventriculaire gauche L_{vet}.

2. Cardiographe à impédance selon la revendication 1, dans lequel le processeur non linéaire (24) produit une valeur compressée de la dérivée de Z par rapport au temps selon la formule suivante :

$$\frac{\widehat{dZn}}{dt} = \frac{\dfrac{\widehat{dZ}}{dt}}{\sqrt{\dfrac{\widehat{dZ}}{dt} / \dfrac{\overline{dZ}}{dt}}}$$

   dans laquelle :

$$\frac{\widehat{dZ}}{dt}$$

   est une dérivée par rapport au temps de Z,

$$\frac{\widehat{dZn}}{dt}$$

   est la dérivée par rapport au temps compressée de Z, et

   $\dfrac{\overline{dZ}}{dt}$ est la valeur normale prédéterminée et est une valeur moyenne de

$$\frac{\widehat{dZ}}{dt}$$

   de pour une population.

3. Cardiographe à impédance selon la revendication 1, dans lequel la valeur normale prédéterminée est essentiellement égale à 1,73 Ohms par seconde.

4. Cardiographe à impédance selon la revendication 1, dans lequel :

   les électrodes (14c, 14d) sont adaptées de manière à être positionnées sur un patient avec une distance de séparation de L ; et
   le cardiographe comporte en outre un dispositif d'entrée pour recevoir (50) les valeurs de la hauteur et de la circonférence et fournissant les deuxième et troisième signaux électriques correspondant H et C indiquant ces valeurs ;

dans lequel le moyen de circuit électrique (24) reçoit les premier, deuxième et troisième signaux électriques et fournit l'indication du volume systolique du patient SV comme une fonction de Z, L, et déduit A par l'approximation suivante

$$\frac{L}{K*H*C}$$

où K est une constante prédéterminée.

5. Cardiographe à impédance selon la revendication 4, dans lequel le moyen de circuit électrique (24) calcule SV selon la formule suivante :

$$SV = \frac{L}{K*H*C} \frac{\rho*C^2*L*L_{vet}}{4*\pi*Z_0} \frac{d\hat{Z}_n}{dt}$$

où :

ρ est la résistivité du sang ;
Z_0 est une impédance ligne de base moyenne ;
L_{VET} est un intervalle d'éjection ventriculaire gauche ; et

$$\frac{d\hat{Z}_n}{dt}$$

est la dérivée par rapport au temps compressée de l'impédance Z.

6. Cardiographe à impédance selon la revendication 1, dans lequel

$$\frac{d\hat{Z}_n}{dt}$$

est l'amplitude de la dérivée minimum par rapport au temps compressée de Z.

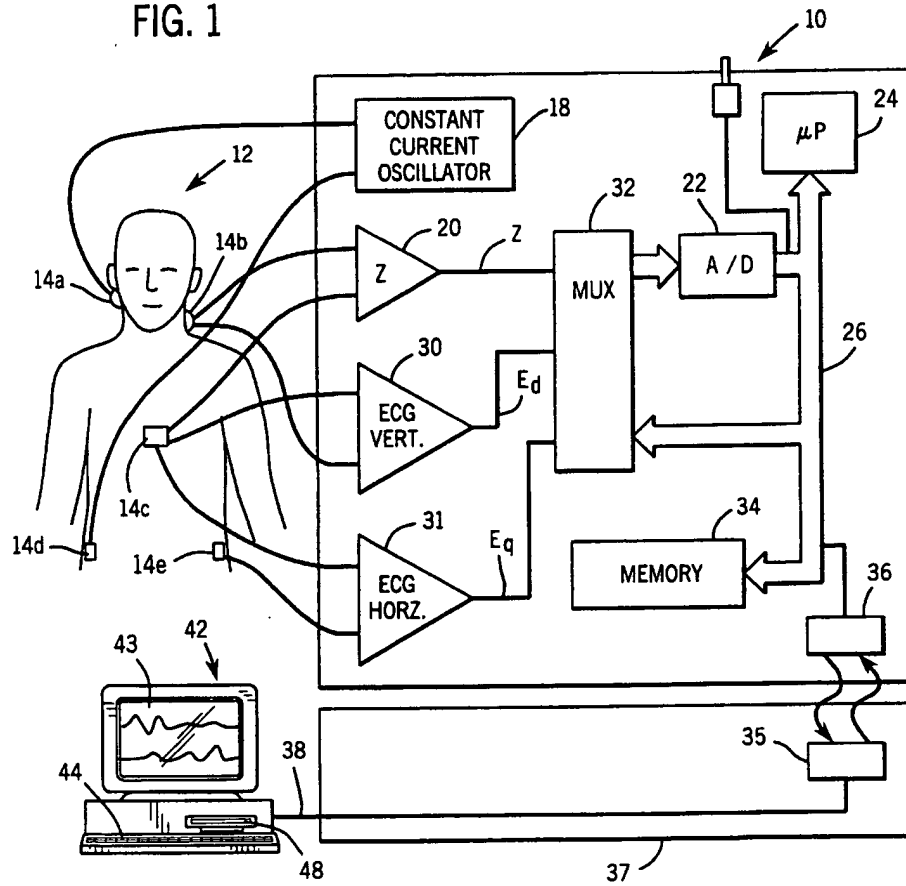7. Cardiographe à impédance selon la revendication 4, dans lequel K est essentiellement égal à 0,875.

FIG. 1

FIG. 2

$$\text{START}$$

INPUT VALUES OF
ρ, C, L, H — 50

START ACQUISITION
OF ECG, Z
DATA — 54

BUILD REF.
ECG WAVE — 58

ENSEMBLE AVE
Z WAVE — 56

CALCULATE
$\dfrac{d\hat{z}}{dt}$ — 59

COMPRESS
$\dfrac{d\hat{z}}{dt}$ TO $\dfrac{d\hat{z}n}{dt}$ — 60

DETERMINE HR
L_{vet}, Z_0 — 52

CALCULATE
SU, CO — 64

DISPLAY
DATA — 66

$$\text{END}$$